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(54) Positron annihilation imaging
device using multiple offset rings of
detectors

(57) A positron annihilation imaging
device comprises two or more coaxial
circular arrays of detectors (2, 2'),
with the detectors in one array
angularly offset with respect to the
detectors in the adjacent array to
detect more than one tomographic
image simultaneously through
different cross-sections of a patient.

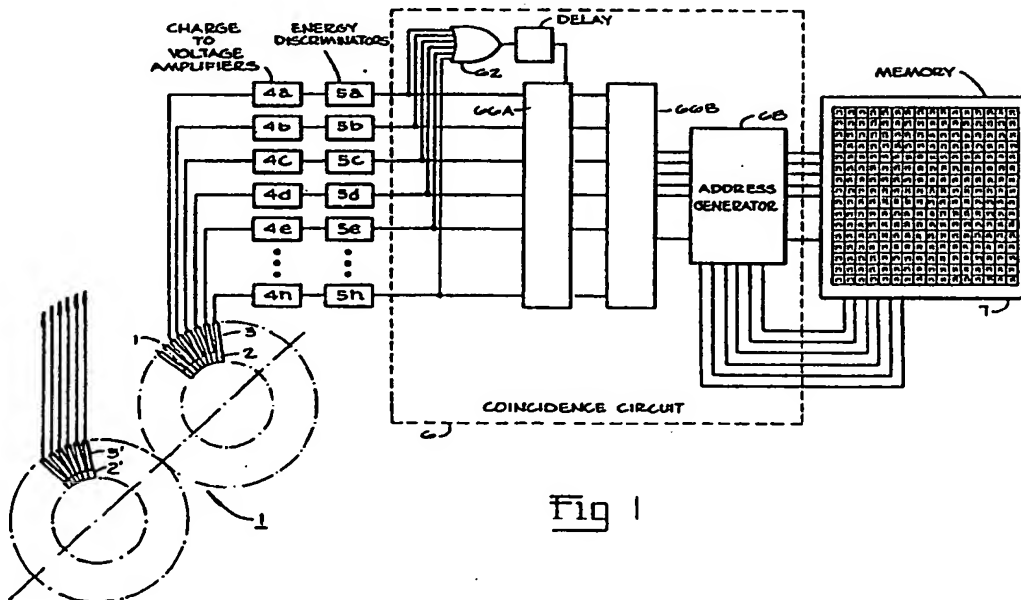


Fig 1

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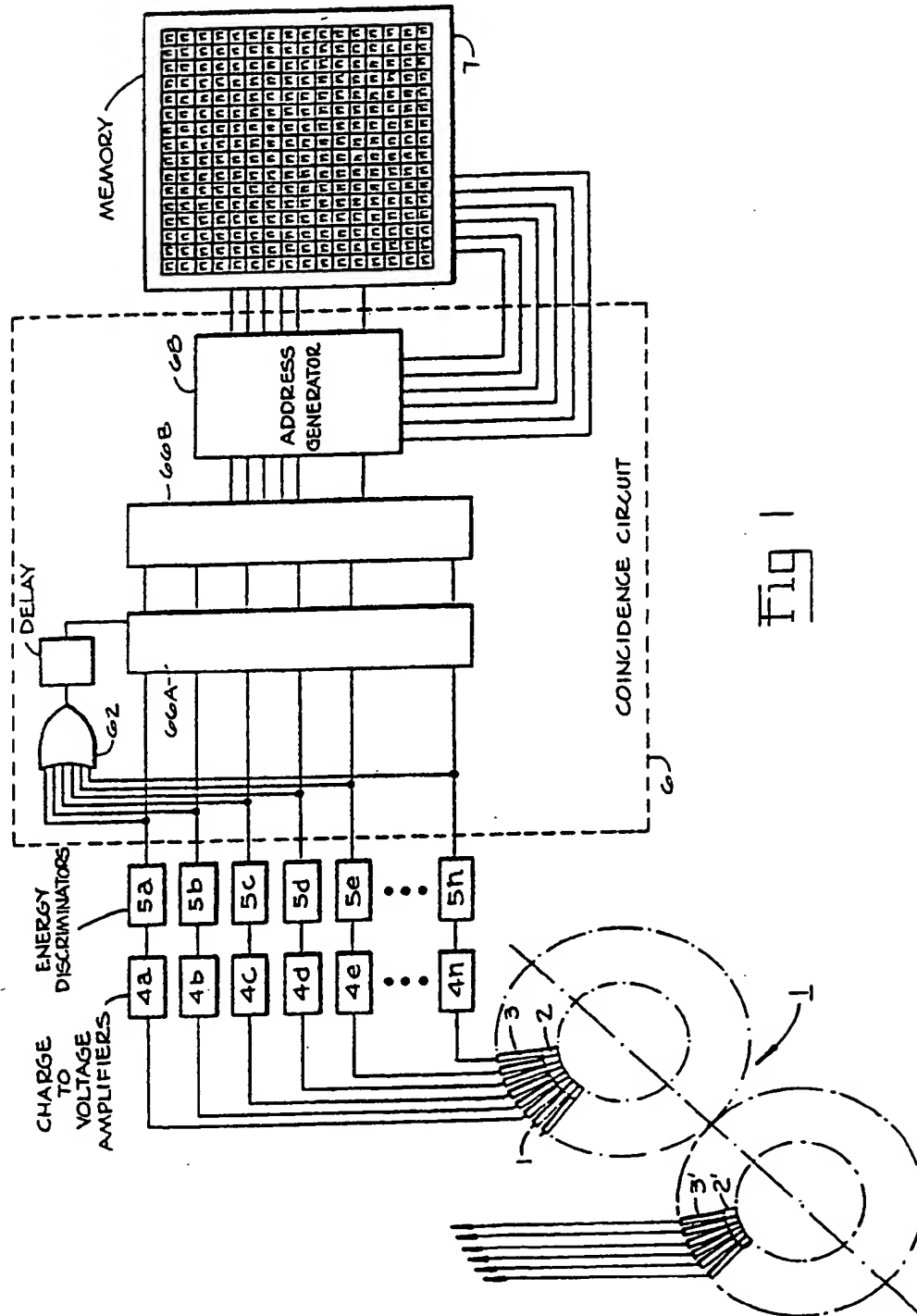
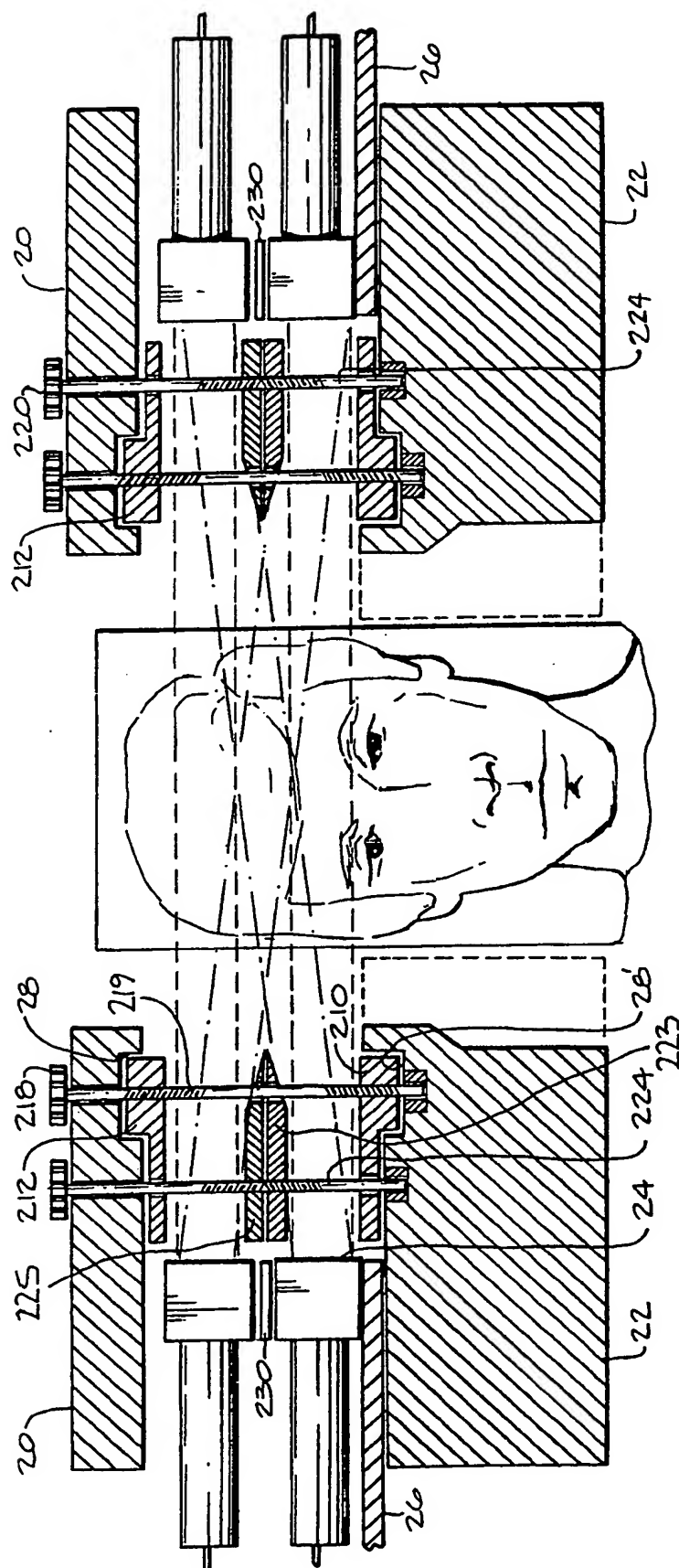


Fig 1



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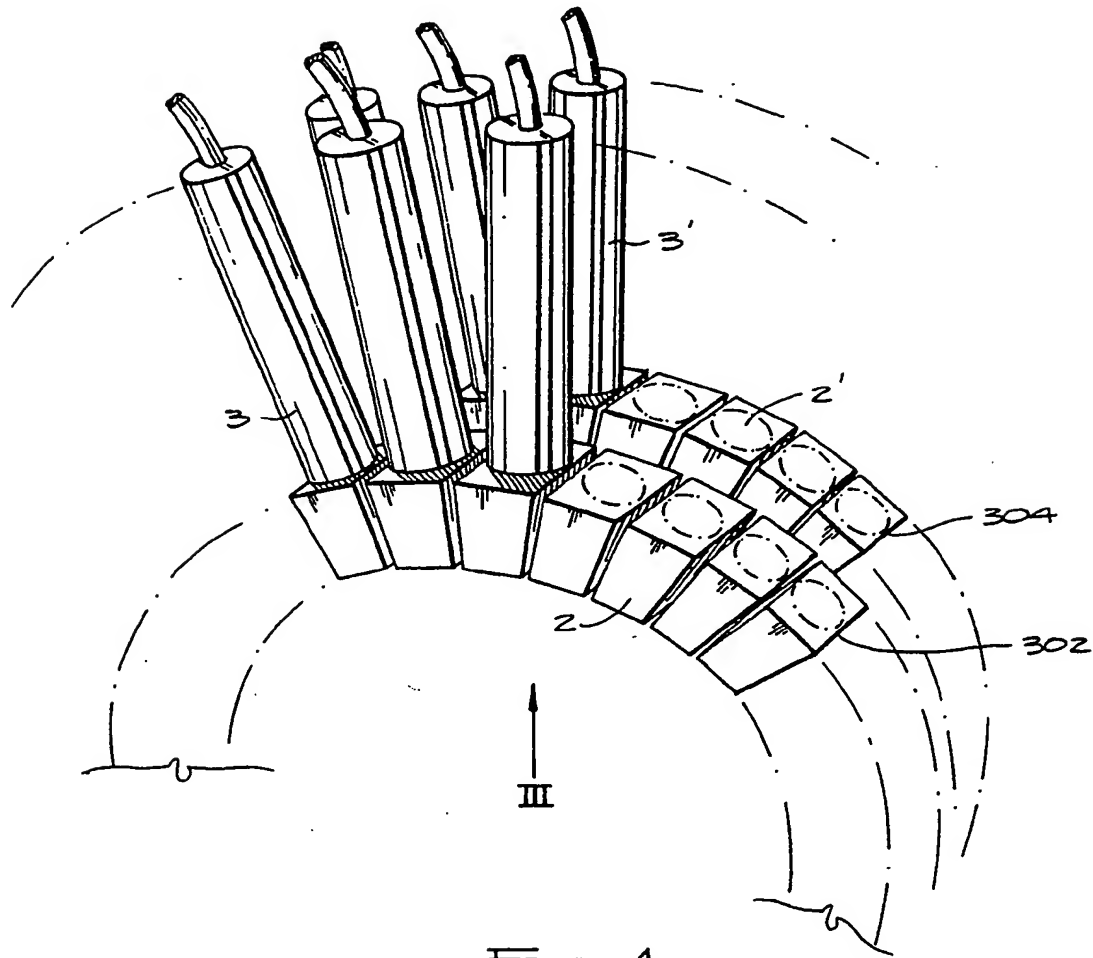


Fig 4

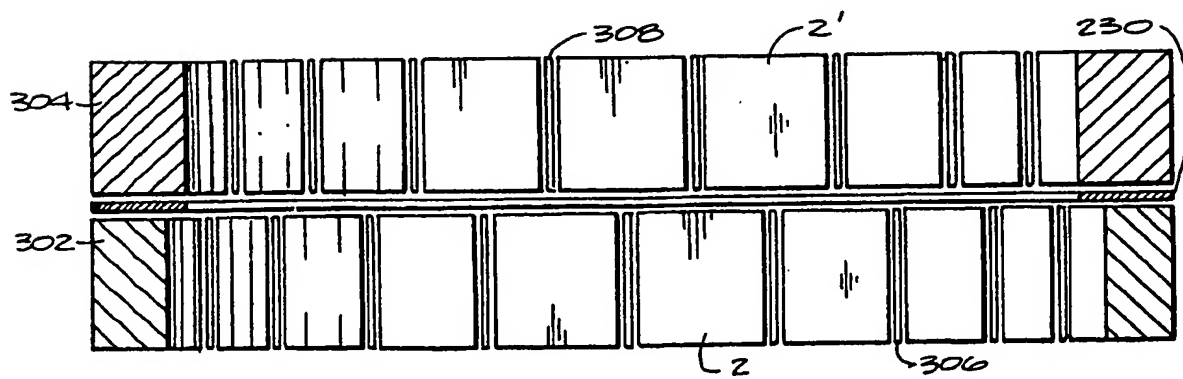


Fig 3

SPECIFICATION

Positron annihilation imaging device using multiple offset rings of detectors

This invention relates generally to positron emission tomography and more particularly to devices which use an array of scintillation detectors to detect the annihilation radiation from positron disintegration and use this information to reconstruct an image of the distribution of positron emitting isotopes within a body.

Positron emission tomography is a technique for measuring the concentration of a positron emitting isotope through a sectional plane through the body. Normally the isotope is used to label a substance which circulates with the blood and may be absorbed in certain tissues. The technique allows the actual concentration in the slice to be determined if the device is suitably calibrated.

Certain isotopes decay by emitting a positively charged particle with the same mass as the electron (positron) and the neutrino from the nucleus. In this process one of the protons in the nucleus becomes a neutron, so that its atomic number goes down while its atomic weight remains constant. This positron is ejected with a kinetic energy of up to 2 MeV depending on the isotope and loses this energy by collisions while travelling a distance of up to a few mms in water. When it has reached thermal energies it interacts with an electron and they mutually annihilate one another. The rest mass of the 2 particles is transformed into 2 gamma rays of 511 keV which are emitted at 180° in the 'center of mass' coordinates of the original particles. The 2 gamma rays may be detected by suitable devices. If these devices measure the energy of the gamma rays at 511 keV and register this energy almost simultaneously it may be assumed that the origin of the radiation is on a straight line between the 2 detectors. Several detectors may be used in an arrangement so that many coincident events may be imaged during the same time interval. Then the information from these detectors is processed by a computer using image reconstruction techniques in order to find the location of distribution of the positron emitting isotope.

A device for imaging positron annihilation radiation comprises the following basic parts:

(1) A number of detectors arranged in a precise geometrical pattern. These detectors are normally scintillation detectors in one or several planes, and these detectors are normally arranged in a polygonal pattern or around the circumference of a circle. Scintillation detectors emit a light flash each time they absorb gamma radiation which may or may not arise from the mutual annihilation of a positron and electron. The intensity of the light flash is proportional to the gamma ray energy.

(2) The device must contain a means of converting the light flash to an electrical charge pulse. Its amplitude is proportional to the light intensity.

(3) The device must contain a means of

determining that the charge pulse could have arisen from a gamma ray whose energy was approximately equivalent to the mass of the electron at rest (511 keV).

(4) The device must have an electric circuit capable of determining that 2 and only 2 detectors each recorded gamma rays or appropriate energy within a short time interval (coincidence resolving time). These detectors are said to have recorded a 'coincident event'.

(5) The device must have an electric circuit which determines which 2 detectors out of the many possible combinations recorded the so-called 'coincident event'.

(6) The device must have a memory in which it can record how often each pair of detectors record a 'coincident event'. The memory may be part of the random access memory of a general purpose computer.

(7) The device is required to use an algorithm through which the information in the memory may be transformed into an image of the distribution of positron annihilation per unit time in a cross-section surrounded by the detectors. The sequence of steps described by this algorithm may be programmed into a general purpose computer.

Accordingly it is a main object of this invention to provide a means for recording more than one tomographic image simultaneously through different cross-sections of a patient.

In accordance with the present invention there is provided a positron annihilation imaging device comprising:

(a) a first array of N detectors equally spaced around a first circle and disposed in a first common plane; and

(b) a second array of N detectors equally spaced around a second circle and disposed in a second common plane, the first and second circles being coaxial, said first and second planes being parallel and spaced apart, the detectors in the first array being offset angularly with respect to said detectors in the second array by an angle substantially equal to $360/2N^\circ$ subtended at the said common axis.

Some embodiments may also include at least one additional array of N detectors equally spaced around and axially circularly disposed in one or more planes spaced from said common planes, the detectors in said additional array, being offset by $360/2N^\circ$ relative to the detectors in an adjacent one of said first or second arrays of detectors.

The invention is described further hereinafter, by way of example, with reference to the accompanying drawings, in which:—

Fig. 1 is an overall block diagram of an apparatus embodying the invention;

Fig. 2 is a cross-sectional view showing how the coincident events are obtained from two rings of detectors to produce three independent tomographic images of consecutive cross-section through a patient;

Fig. 3 shows the detector placement in a two-ring system which is able to produce three cross-

sectional slices; and

Fig. 4 is a pictorial view of two rings of detectors, the detectors in one ring being offset in relation to the detectors in the outer ring.

5 A preferred embodiment of the positron annihilation imaging device 1 is shown in outline form in Fig. 1. The device comprises two or more rings of detectors 2, 2' which surround the object being imaged in two or more planes. The electrical
10 signals from these detectors are amplified at 4a ... 4n, and their energy is measured in energy discriminators 5a ... 5n, the outputs of the energy discriminators being processed by a coincidence circuit 6. The output of the coincident circuit is
15 used to increment memory locations in a general purpose computer. The computer then reconstructs an image of the distribution of the positron imaging isotope in the cross-sections which were scanned.

20 Further details of the circuit of Fig. 1 will be found in our copending U.K. patent application No. 8003837 entitled "Coincidence Analysis Circuit for Positron Annihilation Imaging Device".

25 The preferred embodiment contains two rings of 64 trapezoidal shaped bismuth germanate detectors which are separated by thin tungsten septa. (See Fig.2).

30 For further details of the detector shape and arrangement for a positron annihilation imaging device, reference may be made to our copending U.K. patent application No. 8003836 entitled "Scintillation detector for annihilation radiation in positron disintegration process".

35 These two rings of detectors are rotatable with respect to one another by half the angular separation of the detectors (2.8°).

40 In normal operation of a single slice positron annihilation detector ring it is desirable to rotate the single detector ring in an oscillatory motion to and fro by half the angular separation of the detectors. The purpose of this is to double the
45 number of points in each parallel projection which is used in the image reconstruction technique. The other purpose of this is to uniformly sample the object to be scanned in such a way that the sampling of the projection is done at points which are not greater than half the widths of the detector pair aperture function apart. This is necessary to prevent "aliasing errors" in the reconstructed
50 image.

55 This oscillatory motion takes a finite time (in the preferred embodiment about one third of a second) which limits the rate at which consecutive images can be obtained to approximately two thirds of a second per image. An image can be reconstructed of a 'central slice' between the two planes of detectors when two rings of detectors are used as shown in Fig. 2. Since this central slice is viewed by twice the number of detectors, twice
60 the number of counts per unit time are recorded by the detectors from this area. Because of this, an image can be reconstructed from the center slice in half the time it takes to obtain data of the same statistical accuracy from the outer slices. By
65 offsetting the detector rings by half the angular

separation between detectors, it is seen that data is collected from the same number of points as would be collected from a single ring in its normal and rotated positions. Thus the imaged area is
70 sampled finely enough to allow an image to be reconstructed from a central slice without having to rotate the ring at all. This means in practice that images can be reconstructed from the central slice at a rate determined only by the amount of isotope
75 administered to the patient, the statistical accuracy required in the final image, and the transfer of raw data to more permanent storage (magnetic disc).

Referring now to Fig. 3, there is shown a sectional side elevation of adjacent portions of
80 two rings of detectors, U1 — U64 in one ring and L1 — L64 in the other ring.

The detector rings are separated by a thin tungsten septum 230 which prevents penetration of unwanted radiation from one detector ring to
85 the other. The adjacent detectors in the same row are also separated by respective tungsten septa 306, 308. The device also includes a pair of collimators in the form of annular end plates for preventing radiation outside the slice being
90 viewed to penetrate into one of the detectors. This is shown best in Fig. 2. An outer collimator comprises a first pair of annular lead plates 210, 212 which are displaceable towards and away from each other on rotatable screw-threaded rods
95 219 driven by a motor and belt drive (not shown) and toothed wheels 218. An inner collimator comprises a second pair of annular lead plates 223, 225 which are displaceable towards and away from each other on rotatable screw-
100 threaded rods 224 driven by another motor and belt drive (not shown) and toothed wheels 220. Reference is directed to our concurrently filed U.K. Application No. entitled "Positron
105 annihilation imaging device having movable collimator" for further description of the collimator arrangement.

Figs. 3 and 4 show the positrons of the two rings of detectors 302, 304 with respect to one another. The detectors 302, 304 are angularly
110 separated by half the angle subtended by an individual detector.

In another embodiment a number of detector rings greater than two could be employed in which the detector arrays were alternately staggered by
115 half the angular separation of the detectors so that for N-rings of detectors (2N-1) cross-sectional images can be obtained.

The following advantages are achieved with regard to this invention.

120 More detectors view the radiation emanating from the patient to increase the use of the isotope administered to the patient, then if only one detector array was employed.

125 Encoding and processing data from coincident events involving detectors in adjacent slices allows the whole volume to be imaged eliminating blind spots which would occur if the cross-slice coincident events were not used.

130 Using separate detectors on each slice (as opposed to long detectors and electronically

encoding the location of the event within a detector) allows the count rate capability to be increased with no loss in efficiency. This allows the device to operate over a wider range in count rate.

- 5 By rotating the detectors in adjacent rings by one-half the angular separation of the detectors, the imaging time in an even numbered slice can be improved by a factor of two. This is so since (a)
- 10 twice the number of detector pairs view the centre slice and (b) it is normally necessary to rotate the detector array by half the angular detector separation to achieve the desired spatial resolution and sampling of the imaged plane.
- 15 Other embodiments falling within the scope of the appended claims will occur to those skilled in the art.

CLAIMS

1. A positron annihilation imaging device
- 20 comprising a first array of N detectors equally

spaced around a first circle and disposed in a first common plane; and a second array of N detectors equally spaced around a second circle and disposed in a second common plane, the first and

25 second circles being coaxial, said first and second planes being parallel and spaced apart, the detectors in the first array being angularly offset with respect to said detectors in the second array by an angle substantially equal to $360/2N^\circ$

30 subtended at the said common axis.

2. A positron annihilation imaging device as claimed in claim 1, further including at least one additional array of N detectors equally spaced around and coaxially circularly disposed in one or

35 more planes spaced from said common planes, the detectors in said additional array, being offset by $360/2N^\circ$ relative to the detectors in an adjacent one of said first or second arrays of detectors.

40 3. A positron annihilation imaging device substantially as hereinbefore described with reference to the accompanying drawings.